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13. ABSTRACT (Maximum 200 words) Kevlar helmets provide the soldier with basic ballistic and impact protection. However, the helmet has recently become a mounting platform for devices such as night-vision goggles, drop down displays, weapon-aiming systems, etc. Although designed to enhance soldier performance, these systems increase the mass of the helmet and typically shift the position of the helmet's center of mass forward. The effects of changing the mass properties of the helmet on head and neck forces and moment on neck muscle activity and fatigue are well documented for aviators and soldiers in vehicles. No research to date has been focused on the effects of helmets of varying mass and mass distribution on head and neck forces and moments during a combat foot soldier's physical activities. Physical demands on the combat foot soldier are substantially different than those on aviators or soldiers in vehicles. Therefore, changing the mass properties of the helmet likely has different effects on combat foot soldiers than on aviators. Effects of helmets with and without night vision goggles on head and neck forces and moment during activities such as walking, running, and diving to the ground have never been quantified. The Personal Armor System for Ground Troops (PASGT) helmet has been the standard-issue Army helmet for several years. A medium PASGT helmet weighs 3.54-3.66 pounds, including weight of the shell, chinstrap, suspension, headband, cover, and parachutist pad. The Advanced Combat Helmet (ACH), the next generation helmet, has been the most commonly used helmet of foot soldiers in recent conflicts. It is similar to the PASGT, but weighs a little less. A medium ACH helmet (with chin strap, retention clips, suspension system, and the mounting for the night vision goggles) weighs about 3.11 pounds. Total weight of a medium ACH helmet with PVS14 night vision goggles (weight ~1.2 lbs) is ~4.3 pounds. PASGT offers more ballistic protection than ACH; the ballistic shell is slightly bigger on PASGT than on ACH. The study answers the militarily relevant question of what forces and moments on the head and neck are associated with current used Army helmets and night vision goggles during combat physical activities of the ground soldier.				
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## **USARIEM TECHNICAL REPORT T07-09**

# **THE EFFECTS OF THE PERSONAL ARMOR SYSTEM FOR GROUND TROOPS (PASGT) AND THE ADVANCED COMBAT HELMET (ACH) WITH AND WITHOUT PVS-14 NIGHT VISION GOGGLES (NVG) ON NECK BIOMECHANICS DURING DISMOUNTED SOLDIER MOVEMENTS**

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## Disclaimer

The opinions or assertions contained herein are the private views of the authors and not official views of the Army or Department of Defense. The investigators have adhered to the policies for protection of human subjects as prescribed in Army Regulation 70-25, and the provisions of 32 CFR Part 219.

## BACKGROUND

Kevlar helmets provide the soldier with basic ballistic and impact protection. However, the helmet has recently become a mounting platform for devices such as night-vision goggles, drop down displays, weapon-aiming systems, etc. Although these systems are designed to enhance soldier performance, they increase the mass of the helmet and typically shift the position of the helmet's center of mass forward. The effects of changing the mass properties of the helmet on head and neck forces and moments and on neck muscle activity and fatigue are well documented for aviators and soldiers in vehicles (Ashrafiuon, et al., 1997; Butler & Alem, 1997; Butler, 1992; Alem et al., 1995; Knight and Baber, 2004; Petrofsky and Philips, 1982; Philips and Petrofsky, 1983; Thuresson et al., 2003). No research to date has been focused on the effects of helmets of varying mass and mass distribution on head and neck forces and moments during a combat foot soldier's physical activities. Physical demands on the combat foot soldier are substantially different than those on aviators or soldiers in vehicles. Therefore, changing the mass properties of the helmet likely has different effects on combat foot soldiers than on aviators. The effects of helmets with and without night vision goggles on head and neck forces and moment during activities such as walking, running and diving to the ground have never been quantified.

The Personal Armor System for Ground Troops (PASGT) helmet has been the standard-issue Army helmet for several years. A medium PASGT helmet weighs 3.54-3.66 pounds, including the weight of the shell, chinstrap, suspension, headband, cover, and parachutist pad. The Advanced Combat Helmet (ACH) is the next generation helmet, and has been the most commonly used helmet of foot soldiers in recent conflicts. It is similar to the PASGT, but weighs a little less. A medium ACH helmet (with chin strap, retention clips, suspension system, and the mounting for the night vision goggles) weighs about 3.11 pounds. The total weight of a medium ACH helmet with PVS14 night vision goggles (weight ~1.2 lbs) is ~4.3 pounds. PASGT offers more ballistic protection than ACH; the ballistic shell is slightly bigger on PASGT than on ACH. Additionally, the suspension system for PASGT is comprised of several straps which are designed to provide an offset to keep the helmet away from the head. In contrast, the ACH is designed with comfort pads in the helmet also designed to provide an offset between the helmet shell and the head. However, the comfort pads are additionally designed to increase the stability of the helmet on the head.

In this study, data were obtained on the effects of two currently fielded Army helmets, both with and without modern night-vision goggles, on head and neck forces and moments during treadmill walking, treadmill running and diving to the ground. The study answers the militarily relevant question of what forces and moments on the head and neck are associated with currently used Army helmets and night vision goggles during combat physical activities of the ground soldier. This research was funded through Department of the Army contract W911QY-04-P-0302 in support of Science and Technology Objective (STO) III.Z (MOM.01) Head Supported Mass (HSM): Warfighter Health and Performance.

## EXECUTIVE SUMMARY

A foot soldier's night-vision goggles (NVG) and other helmet-mounted devices increase head-supported mass and move the helmet center of mass forward. This study measured the effects on head and neck biomechanics of wearing an Advanced Combat Helmet (ACH) and the Personal Armor System for Ground Troops (PASGT) helmet with and without NVG during simulated combat activities.

Eleven subjects walked (3.0 mph), ran (6.0 mph), and dove to prone while wearing a foam helmet (control), ACH without NVG, ACH with NVG, PASGT without NVG and PASGT with NVG. Acceleration and kinematic data were obtained from 3 helmet-mounted tri-axial accelerometers and video cameras. Software calculated the forces and moments at the atlas in the head coordinate system. The angular excursion between the helmet and head, based on markers on the face and helmet, was calculated for walking and running, but not for the dive to prone because the markers were not always visible, averaging  $1.0^\circ$  during walking and  $2.1^\circ$  during running.

For walking, the neck compressive forces were closest to expectations, with the control helmet (C) producing lower peak forces ( $55 \pm 1$  N) than either helmet with or without goggles ( $72 \pm 1$  N). The helmets with goggles (HG) produced higher forces ( $73 \pm 1$  N) than the helmets with no goggles (H,  $70 \pm 1$  N). The PASGT produced higher forces ( $72 \pm 1$  N) than the ACH ( $71 \pm 1$  N). The results for peak neck flexion moment were more equivocal. While HG produced a higher mean moment ( $3.7 \pm 0.3$  Nm) than H ( $3.1 \pm 0.2$  Nm) or C ( $2.9 \pm 0.3$  Nm), the moments for H and C did not differ significantly. Also, the difference in moment between ACH and PASGT did not quite reach significance.

For running, neck compressive force was significantly less for C ( $71 \pm 3$  N) than for H ( $89 \pm 3$  N) or HG ( $91 \pm 4$  N), but the latter two did not differ significantly. Neither did force differ between ACH ( $90 \pm 3$  N) and PASGT ( $89 \pm 4$  N). Peak neck flexion moment was significantly greater for ACH ( $4.8 \pm 0.4$  Nm) than PASGT ( $4.0 \pm 0.4$  Nm), even though ACH was lighter, but moment did not differ significantly between C, H, and HG.

For the dive to prone, as with the running, neck compressive force was significantly less for C ( $79 \pm 4$  N) than for H ( $94 \pm 5$  N) or HG ( $97 \pm 9$  N), but the latter two did not differ significantly. Also, anterior force on the neck was higher for HG ( $61 \pm 9$  N) than C ( $43 \pm 8$  N), but did not differ significantly between H ( $56 \pm 9$  N) and C, H and HG, or ACH ( $55 \pm 9$  N) and PASGT ( $63 \pm 9$  N). Peak neck flexion moment was significantly lower for H ( $8.0 \pm 0.6$  Nm) than for HG ( $9.3 \pm 0.7$  Nm) and unexpectedly lower for H than for C ( $9.9 \pm 0.4$  Nm), but did not differ between C and HG. ACH produced higher neck flexion moment ( $9.2 \pm 0.7$  Nm) than PASGT ( $8.1 \pm 0.6$  Nm). The only significant difference in peak neck extension moment was between H ( $1.3 \pm 0.8$  Nm) and HG ( $2.1 \pm 0.9$  Nm).

In conclusion, wearing a ballistic helmet increased the compressive forces on the neck during walking, running, and diving to prone. However, the addition of NVG further increased the compressive forces only for walking, but not for running or diving to prone. Despite being lighter, the ACH produced higher neck flexion moments than the PASGT, most likely because its foam pads did not allow as much relative movement between the helmet and head as did the strap suspension of the PASGT. The highest neck compression acceleration in any of these activities, 1.8 g, was less than 10% of the 20 g reported by the U.S. Air Force as the lower limit for acute neck injury, but may be enough to cause soreness and discomfort over time.

## INTRODUCTION

Journal articles cited in this report were located via several MedLine searches. Technical reports and military laboratory work cited in this report were located via a Defense Technical Information Center (DTIC) search for related technical reports using various combinations of the keywords "Helmet", "Head Mounted Display", and "Head Supported Mass."

Kevlar helmets designed to provide the soldier with basic ballistic protection have become a mounting platform for devices such as night-vision goggles (NVG). Current NVG designs increase the mass of the helmet and shift the helmet's center of mass (COM) anteriorly, which has been associated with increases in vertical forces and flexion moments around the neck and increased activity and fatigue of the posterior neck muscles among aviators during simulated helicopter flights and during functional tasks (Ashrafiuon, et al., 1997; Butler & Alem, 1997; Butler, 1992; Alem et al., 1995; Knight and Baber, 2004; Petrofsky and Philips, 1982; Philips and Petrofsky, 1983; Thuresson et al., 2003). However, no research has focused on the effects of NVG on the forces and moments on and around the atlas during a combat foot soldier's physical activities.

Combat foot soldiers face physical demands that are substantially different than those of aviators or soldiers in vehicles. For instance, researchers interested in the effects of HSM on aviators may be concerned with the effects of whole body vibration and very high accelerations (such as those experienced during crashes), phenomena not prevalent among foot soldiers. In contrast to an aviator, who may spend a great deal of time sitting, foot soldiers spend the majority of their time walking and contending with obstacles in the field. They carry heavy backpacks, and may be required to walk over uneven terrain, hop or jump from one surface to another and step over or crawl under obstacles. Tactical ground-troop movements always include the possibility of contact with the enemy en route or upon arriving at the planned destination (Army, 1990, FM 21-18), requiring rapid transition from walking to a prone posture. The goal of the present research was to quantify the effect of ballistic helmets with and without NVG on neck forces and moments during activities such as walking, running and diving to the ground.

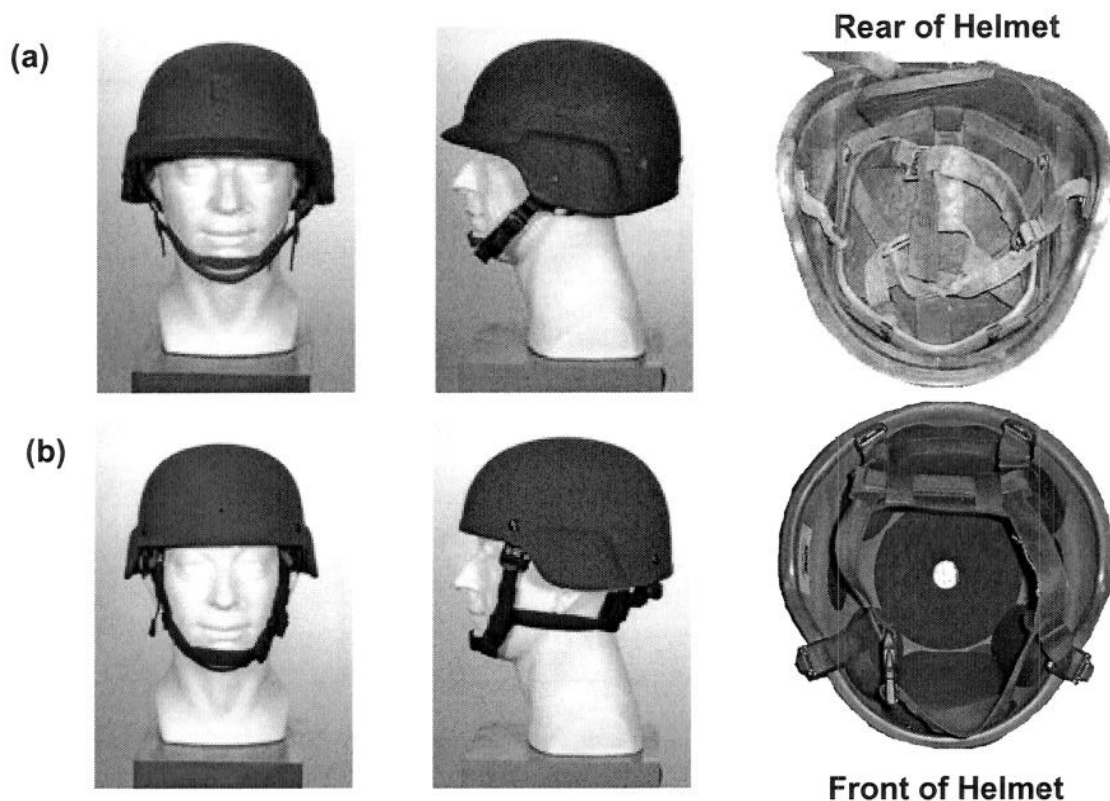
Increasing the mass of the helmet or changing the position of the helmet's center of mass, as when adding night-vision goggles, affects the weight moment of the helmet. Weight moment is defined as the moment caused by the helmet around either the atlanto-occipital complex (AO complex), or the head center of mass (Butler, 1996; Butler 1992). Quantitatively, weight moment is calculated as the product of helmet mass and the horizontal distance between the AO complex and the helmet center of mass. Larger weight moments are associated with increases in forces and moments around the neck, increased activity of the posterior neck muscles (trapezius, sternocleidomastoid (SCM), splenius capitus) and increased neck muscle fatigue among aviators during simulated helicopter flights and during functional tasks (Ashrafiuon, et al., 1997; Butler & Alem, 1997; Knight and Baber, 2004; Phillips & Petrofsky, 1983). Butler (1992), using pilots as subjects, tested aviator helmets weighing over 8.5 pounds (3.9 kg) that had weight

moments up to 170 N·cm (relative to head center of mass) and suggested the optimal weight moment in terms of adding mass to the helmet while minimizing the forces and moments in the neck and posterior neck muscle activity is approximately 83 ( $\pm$  23) N·cm. Butler's work developed guidelines for the use of head supported masses (HSM's) to be worn by the aviator. However, because HSM studies of combat foot soldiers are lacking, there is no information on which to base guidelines on the optimal weight moments for helmets to be worn by foot soldiers.

The Personal Armor System for Ground Troops (PASGT) helmet has been the standard-issue Army helmet for many years. The Advanced Combat Helmet (ACH) is the next generation helmet, and has been used by foot soldiers in more recent conflicts. The most substantial differences between the PASGT and ACH are that the ballistic shell of the PASGT is slightly bigger than that of the ACH and consequently, the PASGT has more mass than the ACH. Additionally, the PASGT has a strap suspension system, while the ACH has comfort pads that increase the stability of the helmet on the head as a suspension system (Figure 1).

Our first hypothesis was that wearing a ballistic helmet while walking, running and diving to prone would result in greater forces and moments on and around the atlas than wearing a lightweight control helmet. Our second hypothesis was that adding NVG to a ballistic helmet would increase the forces and moments exerted on and around the atlas. Our third hypothesis was that the PASGT, being of greater mass than the ACH, would produce greater forces and moments than the ACH on and around the atlas.

Figure 1. PASGT (1a) and ACH (1b) frontal, sagittal and suspension views



## METHODS

All the data were collected at the Center for Military Biomechanics Research, located in the U.S. Army Natick Soldier Research, Development, and Engineering Center in Natick, MA.

Ten healthy male soldiers and one healthy female soldier participated. Only subjects who passed the Army Physical Fitness test within the previous six months, weighed greater than 120 pounds and were between the ages of 18 and 35 were accepted as volunteers. Subjects had no history of back problems or known current injuries or defects to bones or joints, including herniated intervertebral discs. Prior to participation, subjects gave informed consent. The investigators have adhered to the policies for protection of human subjects as prescribed in Army Regulation 70-25, and the provisions of 32 CFR Part 219.

## HELMET MASS PROPERTIES

An anthropometrically correct, ceramic head form representing the 50<sup>th</sup> percentile male was mounted to a test device, which was mounted to a tri-axial force transducer (AMTI Corp, Watertown, MA). The test device was designed to hold the head in specific postures so that COM and moment of inertia (MOI) measurements would be made

coincident with the head anatomical coordinate system (Figure 2). The test device was designed to hold the head in specific postures so that COM and MOI measurements would be made coincident with the head anatomical coordinate system (Figure 3). The offset between the origin of the head coordinate system in the test device and the force transducer coordinate system was measured. The method used to calculate COM was based on the principle that for static objects on horizontal surfaces, a vertical line through the Center of Pressure (COP) passes through the COM of the object, and has been used (with a larger setup) to measure backpack COM in previous USARIEM studies (Harman, et al, 1999, LaFiandra<sup>1</sup> et al., 2003, LaFiandra et al., 2005).

Figure 2. Head coordinate system (Rash et al. 1996)

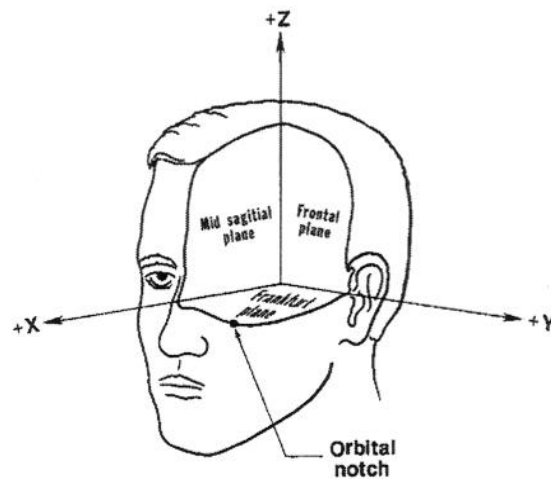
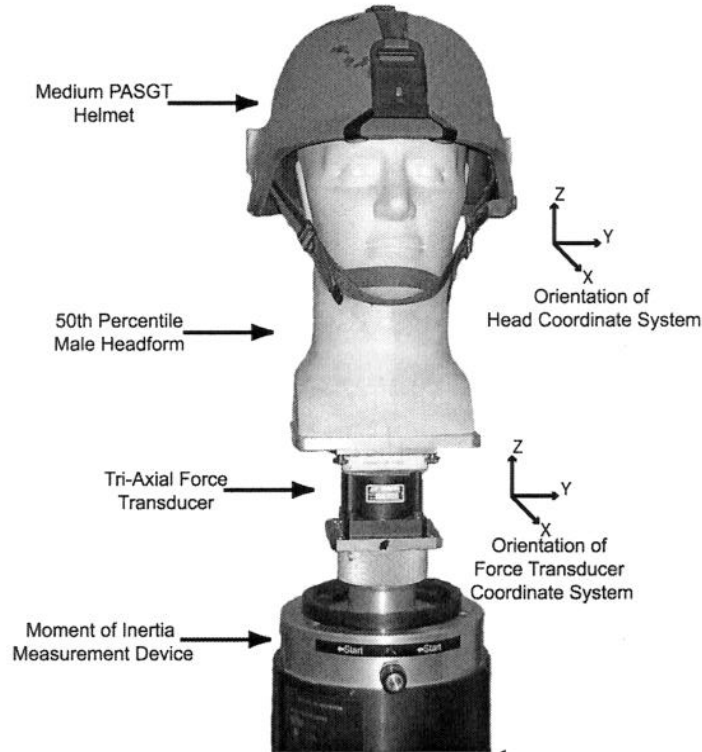


Figure 3. Head/helmet mass properties testing device. In this configuration, the head and force transducer coordinate systems are in the same orientation



The tri-axial force transducer is capable of providing information on the  $F_x$ ,  $F_y$ ,  $F_z$ ,  $M_x$ ,  $M_y$ , and  $M_z$  with respect to the force transducer coordinate system. Raw analog output from the force transducer (in the form of voltage) was amplified by a factory provided amplifier (AMTI MSA- MiniAmp), and then converted to a digital signal by a National Instruments Analog-Digital Card.

The amplified output from the force transducer was converted to forces and moments via factory provided (USARIEM-confirmed) conversion factors. Center of Pressure, (COP) with respect to the force transducer coordinate system, was calculated using equations (Eq. 1 and Eq. 2):

$$\text{Eq. 1: } COP_{x(FT)} = My/Fz$$

$$\text{Eq. 2: } COP_{y(FT)} = Mx/Fz$$

where  $COP_{x(FT)}$  is the x coordinate of the COP with respect to force transducer coordinate system,  $COP_{y(FT)}$  is the y coordinate of the COP with respect to force

transducer coordinate system,  $M_x$  is the moment about the force transducer's x axis,  $M_y$  is the moment about the force transducer's y axis, and  $F_z$  is the vertical force exerted on the force transducer.

In the configuration depicted in Figure 3, the Head/Helmet Mass Properties device can measure the mass of the helmet ( $Mass = F_z / 9.806$ ),  $COP_{x(FT)}$ ,  $COP_{y(FT)}$  and  $MOI_{zz}$ . Because the head mounting bracket was designed to hold the head in specific orientations and the offset between the origins of the head coordinate system and the force transducer coordinate system is known, converting COP with respect to the force transducer to COM with respect to the head coordinate system can be expressed by equation (Eq. 3 and Eq. 4)

$$\text{Eq. 3: } COM_{x(Head)} = COP_{x(FT)} + offset$$

$$\text{Eq. 4: } COM_{y(Head)} = COP_{y(FT)} + offset$$

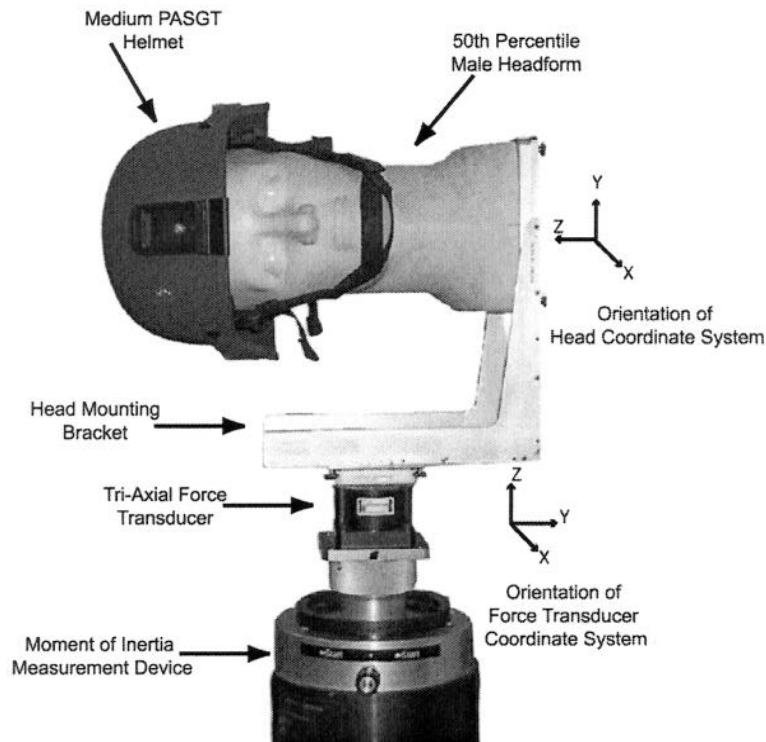
The MOI device actually measures the period of oscillation of the object with respect to the Z axis of the force transducer. The period of oscillation is then used to calculate MOI. As configured in Figure 3, the Z axis of the force transducer is aligned with the Z axis of the head coordinate system. Once the MOI is calculated relative to the Z axis of the head coordinate system, the parallel axis theorem (Eq. 5) is used to determine the MOI around the COM of the head.

$$\text{Eq. 5: } I = I_0 + mx^2$$

where  $I$  represents the moment of inertia about the helmet's COM,  $I_0$  the moment of inertia about the Z axis of force transducer,  $m$  the mass of the helmet and any accoutrements,  $x$  the distance between the COM of the helmet and the measurement axis.

In order to measure  $COP_{z(Head)}$  (and consequently  $COM_{z(Head)}$ ), a special head mounting bracket had to be constructed. The head mounting bracket was designed to hold the head so that the Y axis of the head coordinate system was parallel to the Z axis of the force transducer coordinate system (Figure 4).

Figure 4. COM/MOI measurement device set up to measure  $COM_x$ ,  $COM_z$  and  $I_{yy}$



When configured as in Figure 4, the force transducer can measure  $COP_{x(FT)}$  and  $COP_{y(FT)}$ , which can be converted to  $COM_{x(Head)}$  and  $COM_{z(Head)}$  respectively, by using (Eq. 6 and Eq. 7)

$$\text{Eq. 6: } COM_{x(Head)} = COP_{x(FT)} + offset \text{ (when oriented as in Figure 4).}$$

$$\text{Eq. 7: } COM_{z(Head)} = COP_{y(FT)} + offset$$

When oriented as in Figure 4, the MOI device measured MOI around the Y axis (head coordinate system). MOI around the head's X axis was assumed to be equal the MOI around the Y axis of the head.

Table 1 shows the mass properties of the helmets tested.

Table 1. Helmet mass properties

Helmet	Size	NVG	Mass Kg	COMx (m)	COMy (m)	COMz (m)	Ixx* (Kgm <sup>2</sup> )	Iyy (Kgm <sup>2</sup> )	Izz (Kgm <sup>2</sup> )
ACH-D	Large	N	1.467	-0.012	0.000	0.095	0.014	0.014	0.007
ACH-N	Large	Y	2.010	0.022	0.009	0.119	0.0185	0.025	0.011
ACH-D	Med	N	1.43	-0.007	0.000	0.089	0.013	0.013	0.007
ACH-N	Med	Y	1.972	0.025	0.0086	0.1120	0.0171	0.023	0.011
PASGT-D	Large	N	1.730	0.005	-0.003	0.069	0.013	0.013	0.009
PASGT-N	Large	Y	2.309	0.038	0.006	0.092	0.0179	0.023	0.014
PASGT-D	Med	N	1.609	0.000	0.000	0.082	0.013	0.013	0.008
PASGT-N	Med	Y	2.161	0.032	0.007	0.102	0.0167	0.023	0.012
PASGT-D	Small	N	1.530	-0.002	-0.001	0.084	0.012	0.012	0.007
PASGT-N	Small	Y	2.070	0.029	0.009	0.106	0.0164	0.022	0.011
Control	Med	N	0.600	0	0	0	0	0	0
Control	Large	N	0.600	0	0	0	0	0	0

\*The Helmet MOI device was not designed to measure Ixx. For the No NVG condition, Ixx was set equal to Iyy. For the NVG conditions, Ixx was estimated by determining the contribution of the NVG to MOI for Iyy, then applying the parallel axis theorem to determine Ixx. The COM of the control helmet was assumed to be coincident with the head's COM (therefore was set to (0,0,0)), and considered to be a point mass (therefore the MOI was assumed to be 0 for each axis).

## DATA COLLECTION

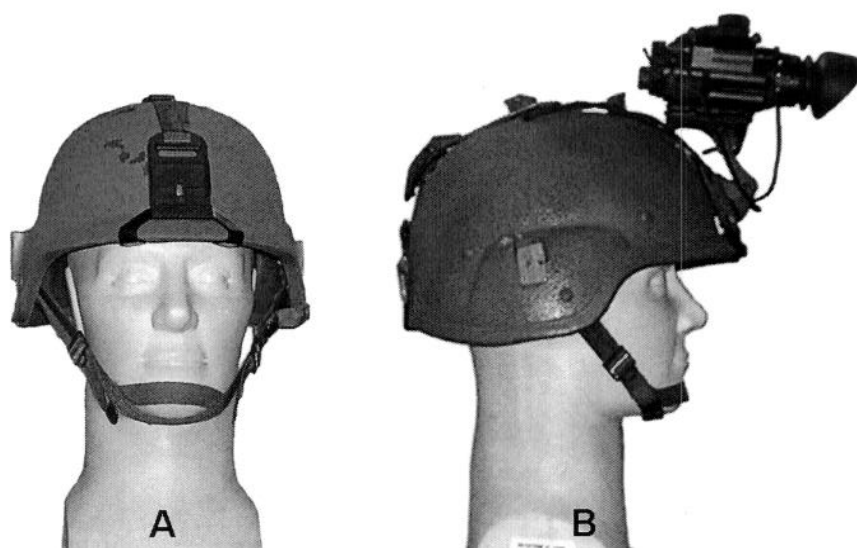
Prior to data collection, anthropometric measurements were made of head height and width, and neck height, circumference and width. Body mass was measured with an electronic scale (SECA Model 770, Germany) and body height was measured using an anthropometer (GPM, Switzerland). Mass properties for each subject's head were estimated based on head height and width, and anthropometric tables from Armstrong (1988).

There were a total of 5 conditions for each task and subject:

1. Control Condition (Lightweight Hockey Helmet)
2. PASGT with no goggles (PASGT-H)
3. PASGT with goggles (PASGT-HG)
4. ACH with no goggles (ACH-H)
5. ACH with goggles (ACH-HG)

Figure 5 shows a PASGT helmet both without and with NVG.

Figure 5. PASGT without (a) and with (b) NVG



The helmets were tested in a balanced order. The total (skin out) weight carried by the subjects was approximately 40 pounds including a simulated fighting vest and mock M16.

An 8 camera Qualisys (Glastonbury, CT) motion capture system was used to capture position data of reflective markers placed bilaterally on bodily landmarks (the 5th metatarsal head, lateral malleolus, lateral femoral condyle, greater trochanter, ulnar styloid, olecranon, acromion process and zygomaticofacial foramen) at 100 Hz. Additional markers were placed halfway between the glabella and nasion (the 'brow'), bilaterally on the transverse process of the atlas, approximated to be 1 cm anteroinferior from the tip of the mastoid process (Moore, 1992), and on the accelerometers on the helmet.

## ACCELEROMETRY

Three Entran (Entran Inc., Fairfield, NJ) accelerometers (Figure 6) were mounted orthogonally to the test helmets (Figure 7). This setup aligned the coordinate systems of each of the accelerometers with the helmet coordinate system, and allowed for the offset between each accelerometer and the helmet coordinate system to be known. In the control condition subjects wore a lightweight hockey helmet to which the accelerometers were mounted. The data from the three accelerometers were to be used in calculating the forces and moments on the atlas joint. Consequently, knowledge of the orientation of the accelerometers, and their positions relative to each other was essential. The three accelerometers were mounted level in the vertical axis and at known distances apart. After the accelerometers were mounted, the distances between them were independently measured for each helmet.

Figure 6. The accelerometer (with a quarter for scale) and the accelerometers mounted on the helmet

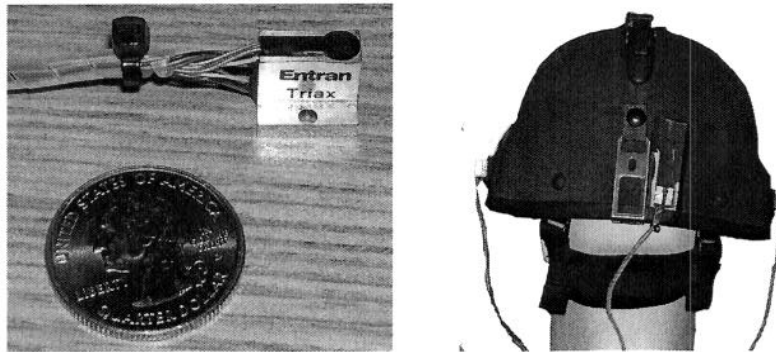
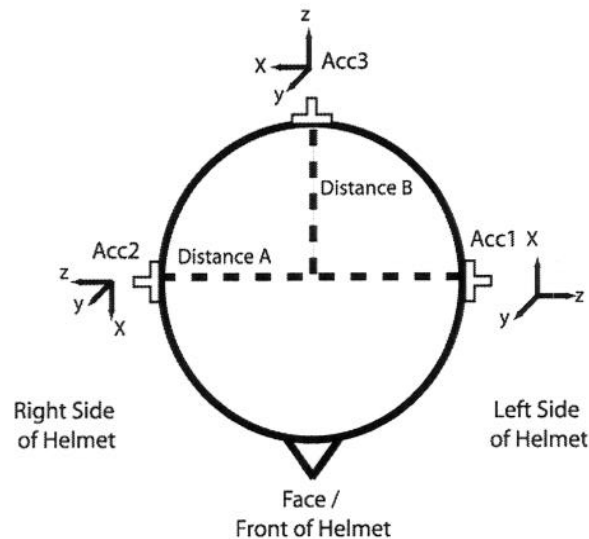


Figure 7. Schematic top view of test helmets indicating accelerometer orientations



Each accelerometer was capable of reading acceleration in three orthogonal planes. Consequently, nine channels of acceleration data were collected. Output from the accelerometers, in the form of voltage, was amplified with Entran amplifiers that were carried by the volunteer in a backpack. The analog output from the amplifiers was sampled at 1000 Hz by a Qualisys AD board in the same computer on which the kinematic data were collected. This equipment setup ensured that the kinematic and accelerometer data were time-synchronized.

## **PREPARATION OF VOLUNTEERS FOR TESTING**

Subjects reported for two nonconsecutive days of data collection wearing shorts, a T-shirt, combat boots and socks. One day was dedicated to collecting the dive to prone data; the other day was dedicated to collecting walking and running data. On the first day of data collection, laboratory procedures were explained to the subjects, anthropometric measurements were taken, and the correct helmet size was determined.

Upon arrival for data collection, the subjects were briefed on the tasks for that day. A researcher then prepared the subjects for data collection by putting the reflective markers on the previously listed landmarks. Because the straps holding the helmet on the head interfered with the positioning of the atlas markers, a method for estimating the position of the atlas similar to the bite-bar method used by Butler (1992) was used. However, the nature of our tasks (diving to the ground) made using a bite bar potentially unsafe. Once the subjects were prepared for data collection by having all of the kinematic markers placed on body landmarks (including markers on the atlas), but before the helmet was put on, a 5 second 'Static Atlas Data Collection' was taken. These data were used to determine the relationship between the markers on each side of the atlas, the brow and each zygomaticofacial foramen. Once this relationship was determined, under the assumption that the head was a rigid body, the position of the atlas was determined without markers on the atlas. The atlas markers were then removed to properly fit the helmet.

Once the Static Atlas Data Collection was finished, a researcher helped the subjects don the simulated fighting load, including the helmet instrumented with the accelerometers. If needed, the subjects were allowed to practice that day's specific task before data collection.

## **SOLDIER ACTIVITIES TESTED**

### **Dive to Prone**

After being prepared for data collection, the volunteer donned the first helmet and prepared for the dive to prone. All the subjects were previously trained (by the Army) on how to complete a dive to prone. With the subject in a standing position, a data collector started collecting kinematic and acceleration data. After 3 seconds of the subject maintaining a static position, the data collector made an auditory signal, after which the subject quickly executed a dive to the prone position onto a gymnastics mat. The 2-3 seconds of maintaining a static position before the dive was the *static helmet* portion of the data collection trial. The subject maintained the prone posture until the end of the data collection trial. Total data collection time per trial was 10 seconds. Subjects would repeat the procedure until 3 data collection trials were collected for that helmet condition. The procedure was repeated for each of the five helmet conditions for a total of fifteen acceptable data collection trials per session.

## **Walking and Running**

After being prepared for data collection, the volunteer donned the first helmet and mounted the treadmill. The subject was asked to look straight ahead while a 5 second *static helmet* collection of kinematic and acceleration data was taken. The treadmill belt speed was brought up to 3.0 miles per hour at zero grade. After a 5 minute adaptation/warmup period, kinematic and acceleration data were collected for 10 seconds. The subject was then allowed to rest for up to 5 minutes, after which the subject remounted the treadmill and the belt speed was brought up to 6.0 miles per hour at zero grade. After a 2-5 minute adaptation period, 10 seconds of kinematic and acceleration data were collected. At the conclusion of a running data collection trial, a five minute rest was allowed. During this rest period, the next helmet was prepared. The same procedure, including the *static helmet* data collection, was repeated for each of the 5 helmets.

## **DATA ANALYSIS**

### **Accelerometry**

Each of the three accelerometers provided 3 channels of acceleration data, corresponding to accelerations along each accelerometer's x, y, and z axes, for a total of 9 channels of acceleration data. Raw data from each accelerometer were amplified with factory provided amplifiers carried in a backpack by the volunteer. These amplifiers were connected via wires to the analog input of a computer. The voltages at each input channel were converted at the rate of 1000 Hz to digital values and stored in computer data files. Factory-provided calibration factors were used to convert the raw data into actual accelerations.

Data from the *static helmet* portion of the data collection were used to zero each accelerometer for that specific trial. Zeroing the accelerometers was based on the voltages read from the accelerometers and accounted for the orientation of the accelerometers with respect to gravity (measured by helmet orientation). The helmet was considered to be static when there were 5 consecutive frames of kinematic data in which each helmet marker moved less than a total of 1.5 mm. Data from this 5 second window were used to zero the accelerometers.

After zeroing the accelerometers, the raw voltages were converted to accelerations via factory provided calibration factors. Next, each trial of accelerometry data was reviewed by graphing each channel. Because all three accelerometers were attached to the helmet, it was expected that each would experience relatively similar acceleration profiles. Trials with bad data were identified as those in which the output of one or more accelerometer showed spikes where the two other accelerometers did not. Such trials were not analyzed further. Each subsection of the results section details the amount of data used for that part of the analysis.

## **Forces Exerted on and Moments Around the Atlas**

Because this was the first head supported mass study conducted at USARIEM, software specific for head/neck analysis had to be written. Custom-written software determined the forces exerted by the head and helmet on the atlas in three orthogonal directions, in both the world and head coordinate system based on the mass properties of the head and helmet, the accelerometry data, and the kinematic data. Additionally, the software calculated the moments around the 3 axes ( $M_x$ ,  $M_y$ ,  $M_z$ ) of the Atlas.

The analysis is based on the assumption that the head and helmet are combined as a rigid body system. Equations of motion (Eq. 8 and Eq. 9) that relate changes in linear momentum to external force applied, and changes in angular momentum to external moments were used for these calculations.

$$\text{Eq. 8: } \frac{d\vec{P}}{dt} = \frac{d(m\dot{\vec{r}})}{dt} = \vec{F}$$

$$\text{Eq. 9: } \frac{d\vec{L}}{dt} = \frac{d(I\vec{\omega})}{dt} = \vec{\tau}$$

where  $\vec{F}$  represents external forces,  $\vec{P}$  linear momentum,  $m$  mass,  $\dot{\vec{r}}$  linear velocity of head and helmet COM,  $\vec{L}$  angular momentum,  $I$  Moment of inertia of the head and helmet,  $\vec{\omega}$ , angular velocity about the head and helmet COM,  $\vec{\tau}$  is the moment due to external forces.

These equations of motion can be written for any arbitrary point, using the transformation (Eq. 10).

$$\text{Eq. 10: } \vec{r} = \vec{R} + C_{WH} \vec{S}_H$$

where  $\vec{r}$  represents the position of the origin in world coordinates,  $\vec{R}$  the position of the origin on the head in world coordinates,  $\vec{S}_H$  is a vector from the head origin to the head COM in the head coordinate system,  $C_{WH}$  the transformation matrix from the head to world coordinates. For the purpose of our analysis, we wrote these equations to calculate forces and moments about the Atlas. For walking, running and jumping from a platform, the position of the atlas for each frame of data was determined as described in the "Subject Tasks" section of the report. Because of missing face marker data during dive to prone, position of the atlas was determined using helmet marker data instead of face marker data. The relationship between the helmet markers and face markers during the static portion of the data collection trial for dive to prone was used to determine the relationship between the helmet markers and the virtual atlas marker.

While all the forces presented in this report are in the head coordinate system, the above step is necessary because each accelerometer reports acceleration in its own coordinate system. The above transformation allows for all accelerometer readings to be expressed in a common coordinate system. Because the orientation of the head coordinate system with respect to the world coordinate system is known, the transformation of forces and accelerations between these two coordinate systems can be readily accomplished.

For one accelerometer, the equations of motion including the transformation to the Atlas become (Eq. 11 and Eq. 12)

$$\text{Eq. 11: } m \left[ \ddot{\vec{R}} + C_{WH} \tilde{\omega}_H \tilde{\omega}_H \vec{S} + C_{WH} \dot{\tilde{\omega}}_H \vec{S}_H \right] = \vec{F}$$

$$\text{Eq. 12: } m \left[ \tilde{S}_H C_{HW} \ddot{\vec{R}} \right] + \tilde{\omega}_H I_H \tilde{\omega}_H + I_H \dot{\tilde{\omega}}_H = \vec{M}_N + \tilde{L}_N C_{HW} \vec{F}$$

where  $\ddot{\vec{R}}$  represents the contribution to the total force due to linear acceleration,  $C_{WH} \tilde{\omega}_H \tilde{\omega}_H \vec{S}$  the contribution due to centripetal acceleration,  $C_{WH} \dot{\tilde{\omega}}_H \vec{S}_H$  the contribution due to angular acceleration,  $m \left[ \tilde{S}_H C_{HW} \ddot{\vec{R}} \right]$  the moment of linear acceleration,  $\tilde{\omega}_H I_H \tilde{\omega}_H$  the gyroscopic moment,  $I_H \dot{\tilde{\omega}}_H$  the moment of angular acceleration, and  $\vec{M}_N + \tilde{L}_N C_{HW} \vec{F}$  the net external moment. Similar equations of motion can be written for each of the accelerometers, providing sufficient data to calculate forces and moments at and around the Atlas joint.

Normalizing force by total weight produces a dimensionless value that represents acceleration in terms of gravity (g; Eq. 13).

$$\text{Eq. 13: } Fi_{normalized} = Fi_{Atlas} / totalweight$$

where  $i$  represents  $x$  or  $z$  and total weight represents the weight of the head + helmet (with accelerometers) + NVG (if applicable).

### **Helmet Motion**

To verify our assumption that the helmet and head act as a single rigid body, the angle between the markers on the face and the markers on the helmet was calculated. Changes in this angle indicate relative movement of the helmet and head. Because of missing face data for the dive to prone, this analysis could not be performed on that part of the dataset. During walking the average range of movement of the helmet (across all subjects and conditions) was 1.0°. During running the average was 2.1°. There are two

sources for this movement: slipping of the skin relative to the skull and movement of the helmet relative to the skin. In the current study, there was no independent measure of these two sources of movement.

### **Data Summary**

For the walking and running, individual strides of data (based on time of peak hip flexion) were extracted as individual trials for analysis. For the dive to prone data, each dive was treated as a single trial. Peak force ( $F_z$ ) and peak moment ( $M_y$ ) exerted by the helmet and head on the atlas and Peak  $F_z$ /(weight of head + helmet and any affixed equipment: totalweight) were determined for each trial of walking, running and dive to prone. Additionally Peak force ( $F_x$ ) and Peak  $F_x$ /totalweight were determined for each trial of dive to prone.

### **Statistics**

A one-by-three analysis of variance (ANOVA) with repeated measures and one within subject effect (helmet) were used to test for differences in all the dependent measures. There were three levels of helmet for this analysis (control, H and HG). When a significant main effect of helmet condition was found, individual ANOVAs between conditions were used to determine specifically which conditions were different.

Additionally, focused analyses were performed. A two-by-two ANOVA with repeated measures and 2 within subject effects (Helmet type and NVG condition) was used to test for differences in all of the dependent variables. There were 2 levels of helmet type: ACH and PASGT, and 2 levels of NVG condition: H and HG. Main effects of helmet type and NVG condition were investigated as well as Helmet x NVG interaction. The control condition was not included in this part of the analysis.

## RESULTS

All forces and moments are expressed in the head coordinate system. The forces and moments exerted on the Atlas were calculated for a total of 122 trials of walking data. There were a total of 27 trials for the control condition, 19 for ACH-H, 24 for ACH HG, 27 for PASGT H, and 25 for PASGT HG. Sample time series of force data (Figure 8) and moment data (Figure 9) for 1 stride of walking in each condition are presented.

Figure 8. Representative walking time series of compressive forces for one stride of data for one subject ( $F_z$  in the head coordinate system)

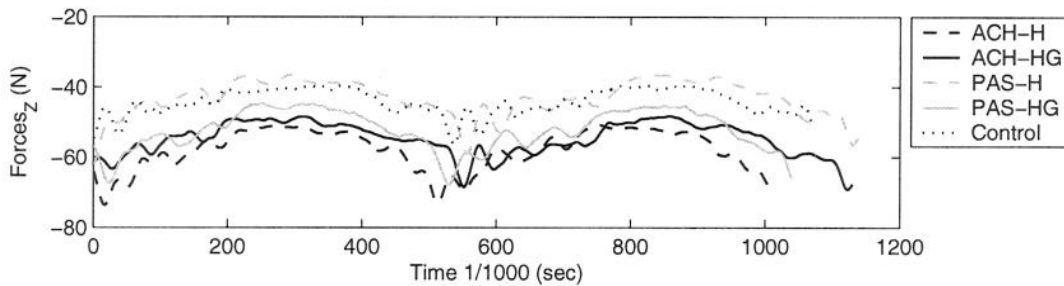
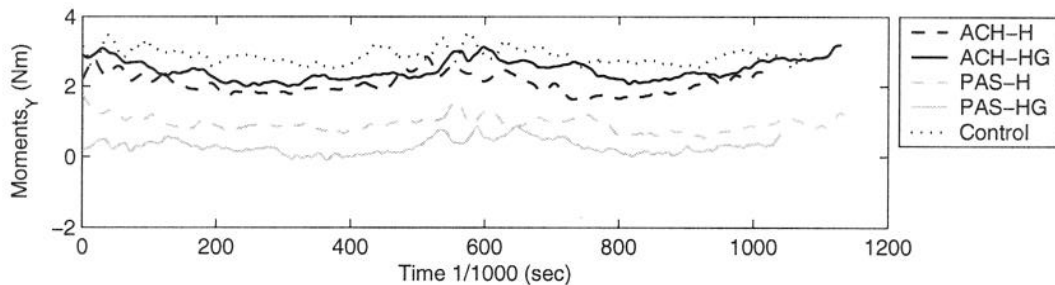


Figure 9. Representative walking time series of neck flexion moment for one stride of data for one subject ( $M_y$  in the head coordinate system)



The forces and moments exerted on the Atlas were calculated for a total of 106 trials of running data. There were a total of 24 trials for the control condition, 21 for ACH-H, 21 for ACH-HG, 21 for PASGT-H, and 19 for PASGT-HG. Sample time series of force data (Figure 10) and moment data (Figure 11) for 1 stride of running in each condition are presented.

Figure 10. Representative running time series of compressive forces for one stride of data for one subject ( $F_z$  in the head coordinate system)

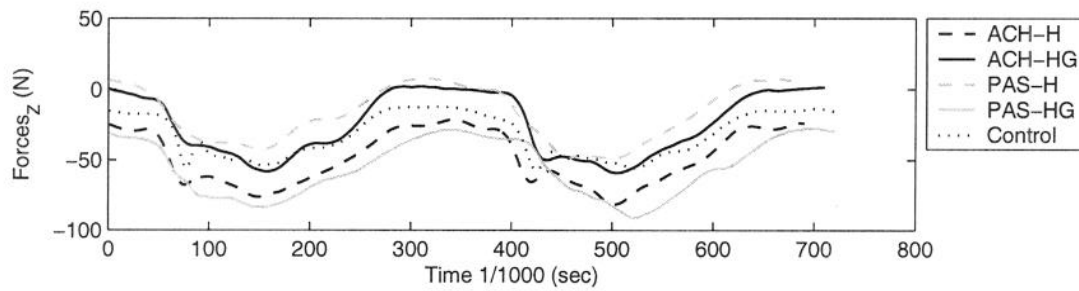
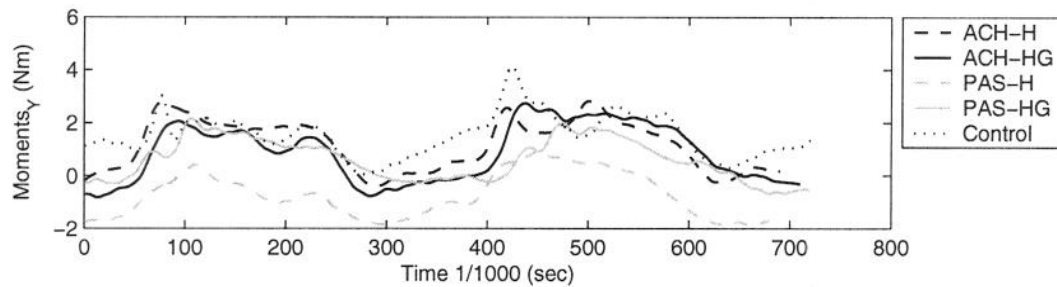


Figure 11. Representative running time series of neck flexion moment for one stride of data for one subject ( $M_y$  in the head coordinate system)



The forces (Figure 12) and moments (Figure 13) exerted on the Atlas were calculated for a total of 91 trials of dive to prone data. There were a total of 20 trials for the control condition, 14 for ACH-H, 15 for ACH-HG, 21 for PASGT-H, and 21 for PASGT-HG. Sample time series of force data for 1 dive to prone trial in each condition are presented.

Figure 12. Representative dive to prone time series of compressive forces and anterior/posterior forces for one dive for one subject (Fz and Fx in the head coordinate system)

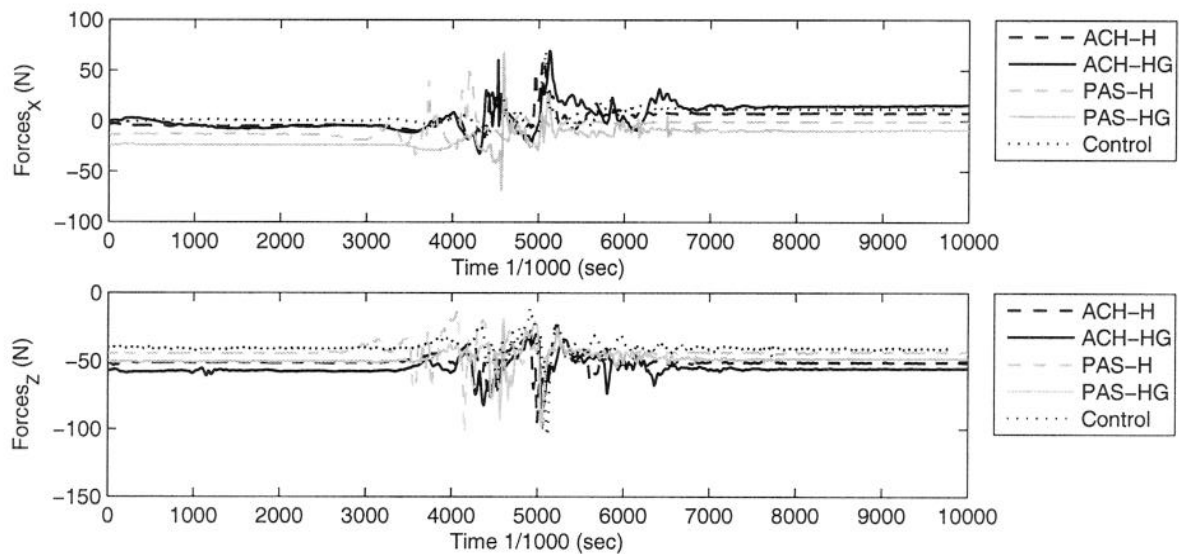
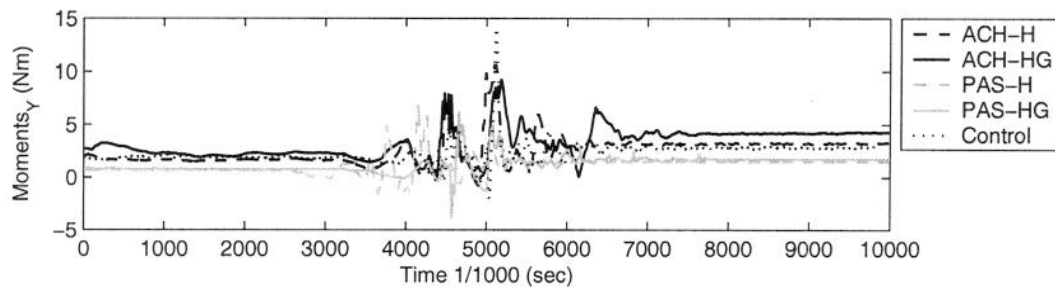


Figure 13. Representative dive to prone time series of neck flexion moment for one dive for one subject (My in the head coordinate system)



There was a significant main effect of helmet configuration (control vs. H vs. HG) on peak Fz during walking, running and diving to prone (Table 2). Tables 3, 4 and 5 summarize the main effects of helmet design, NVG condition and the results of the focused analysis for walking. Tables 6, 7 and 8 summarize the main effects of helmet design, NVG condition and the results of the focused analysis for running. Tables 9, 10 and 11 summarize the main effects of helmet design, NVG condition and the results of the focused analysis for diving to prone. Post-hoc analysis revealed walking, running, or diving to prone in the control condition resulted in lower peak Fz values than observed in H or HG (Tables 3, 4 and 5); these differences were statistically significant. Additionally, post-hoc analysis revealed walking with a helmet in the H configuration resulted in lower peak Fz than walking in the HG condition (Figure 14). There was a significant main effect of helmet condition on peak Fx during dive to prone; post-hoc analysis showed that completing dive to prone in the control condition resulted in a lower peak Fx than in the HG condition.

Table 2. P-Value results from ANOVA and focused analyses

	Variable	Main Effect of Helmet Configuration	Main Effect of Design (ACH vs PASGT)	Main Effect of NVG (H vs HG)	Design x NVG Interaction
Walking	Peak Fz	<b>0.0001</b> <sup>A B C</sup>	<b>0.0485</b>	<b>0.0077</b>	0.1050
	Peak Fz / Totalweight	0.0803	0.7717	<b>0.009</b>	0.0966
	Peak Neck Flexion Moment	<b>0.0425</b> <sup>B C</sup>	0.065	0.0707	0.2763
Running	Peak Fz	<b>0.0001</b> <sup>A B</sup>	0.6588	0.4397	0.5805
	Peak Fz / Totalweight	0.4083	0.7923	<b>0.0336</b>	0.6208
	Peak Neck Flexion Moment	0.1335	<b>0.0008</b>	0.1194	0.5573
Dive to Prone	Peak Fx	<b>0.0087</b> <sup>B</sup>	0.1856	0.4417	0.8451
	Peak Fx / Totalweight	0.6387	0.2267	0.9648	0.8464
	Peak Fz	<b>0.0131</b> <sup>A B</sup>	0.6253	0.6557	0.0643
	Peak Fz / Totalweight	0.4566	0.454	0.3363	0.0545
	Peak Neck Flexion Moment	<b>0.0203</b> <sup>A C</sup>	<b>0.0467</b>	<b>0.0271</b>	0.9021
	Peak Neck Extension Moment	0.0982	0.4169	<b>0.0122</b>	0.9999

Bold indicates  $p < 0.05$

A indicates statistically significant differences between Control and H

B indicates statistically significant differences between Control and HG

C indicates statistically significant differences between H and HG

Table 3. Means (standard deviations) for dependent variables during walking

Variable	Control	ACH-H	PASGT-H	ACH-HG	PASGT-HG
Peak Fz (N)	-55.266 (1.261)	-69.135 (0.914)	-70.933 (1.737)	-73.223 (1.319)	-73.519 (1.375)
Peak Fz / Totalweight	-1.204 (0.034)	-1.266 (0.014)	-1.275 (0.026)	-1.226 (0.016)	-1.205 (0.019)
Peak Neck Flexion Moment (Nm)	2.881 (0.289)	3.370 (0.225)	2.936 (0.273)	4.021 (0.417)	3.454 (0.295)

Table 4. Means (standard deviations) for dependent variables during walking –  
Main Effect of Helmet Design

Variable	Control	ACH	PASGT
Peak Fz (N)	-55.266 (1.261)	-71.32 (0.96)	-72.23 (1.12)
Peak Fz / Totalweight	-1.204 (0.034)	-1.25 (0.01)	-1.24 (0.02)
Peak Neck Flexion Moment (Nm)	2.881 (0.289)	3.72 (0.25)	3.19 (0.21)

Table 5. Means (standard deviations) for dependent variables during walking –  
Main Effect of NVG Condition

Variable	Control	Helmet	Helmet + NVG
Peak Fz (N)	-55.266 (1.261)	-70.15 (1.05)	-73.38 (0.93)
Peak Fz / Totalweight	-1.204 (0.034)	-1.27 (0.02)	-1.21 (0.01)
Peak Neck Flexion Moment (Nm)	2.881 (0.289)	3.13 (0.19)	3.72 (0.25)

Table 6. Means (standard deviations) for dependent variables during running

Variable	Control	ACH-H	PASGT-H	ACH-HG	PASGT-HG
Peak Fz (N)	-71.387 (3.271)	-88.567 (2.801)	-88.526 (3.722)	-91.138 (3.614)	-89.95 (3.642)
Peak Fz / Totalweight	-1.550 (0.077)	-1.624 (0.050)	-1.593 (0.074)	-1.521 (0.049)	-1.473 (0.056)
Peak Neck Flexion Moment (Nm)	3.557 (0.576)	4.411 (0.294)	3.839 (0.459)	5.186 (0.489)	4.199 (0.442)

Table 7. Means (standard deviations) for dependent variables during running –  
Main Effect of Helmet Design

Variable	Control	ACH	PASGT
Peak Fz (N)	-71.387 (3.271)	-89.85 (2.23)	-89.24 (2.51)
Peak Fz / Totalweight	-1.550 (0.077)	-1.57 (0.04)	-1.53 (0.05)
Peak Neck Flexion Moment (Nm)	3.557 (0.576)	4.80 (0.29)	4.02 (0.31)

Table 8. Means (standard deviations) for dependent variables during running –  
Main Effect of NVG Condition

Variable	Control	Helmet	Helmet + NVG
Peak Fz (N)	-71.387 (3.271)	-88.55 (2.24)	-90.54 (2.47)
Peak Fz / Totalweight	-1.550 (0.077)	-1.61 (0.04)	-1.50 (0.04)
Peak Neck Flexion Moment (Nm)	3.557 (0.576)	4.13 (0.27)	4.69 (0.34)

Table 9. Means (standard deviations) for dependent variables during dive to prone

Variable	Control	ACH-H	PASGT-H	ACH-HG	PASGT-HG
Peak Fx Direction (N)	38.907 (24.613)	52.612 (25.506)	57.461 (17.022)	56.466 (27.465)	63.526 (30.98)
Peak Fx / Totalweight	0.843 (0.543)	0.972 (0.481)	0.97 (0.293)	1.017 (0.486)	1.043 (0.516)
Peak Fz Direction (N)	-79.751 (22.548)	-95.104 (12.845)	-92.789 (25.231)	-94.75 (15.34)	-98.661 (20.662)
Peak Fz / Totalweight	-1.731 (0.514)	-1.761 (0.274)	-1.565 (0.43)	-1.713 (0.301)	-1.618 (0.349)
Peak Neck Flexion Moment (Nm)	9.878 (2.729)	8.665 (1.973)	9.787 (2.361)	7.549 (1.633)	8.734 (2.345)
Peak Neck Extension Moment (Nm)	-2.957 (3.148)	-0.875 (0.783)	-1.709 (1.808)	-1.706 (2.42)	-2.43 (2.078)

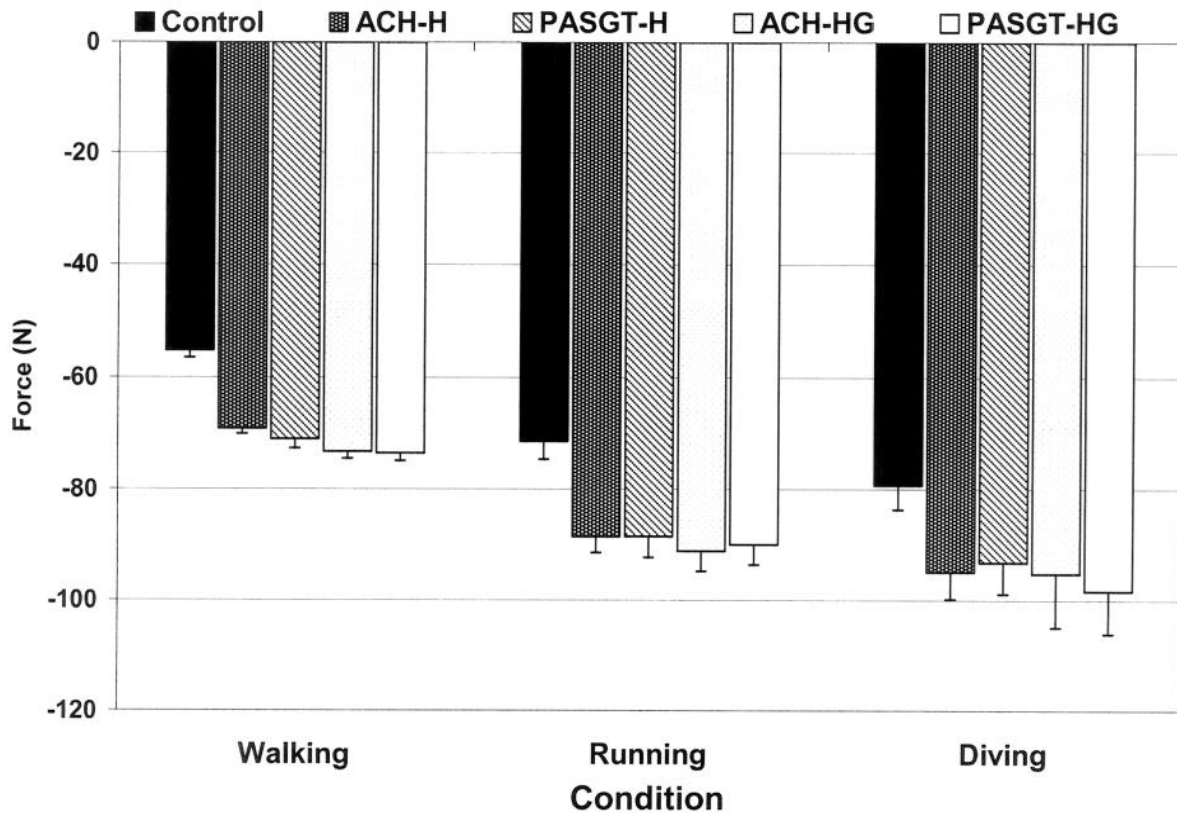
Table 10. Means (standard deviations) for dependent variables during dive to prone –  
Main Effect of Design Condition

Variable	Control	ACH	PASGT
Peak Fx Direction (N)	38.907 (24.613)	54.23 (5.77)	62.73 (6.61)
Peak Fx / Totalweight	0.843 (0.543)	0.95 (0.11)	1.08 (0.11)
Peak Fz Direction (N)	-79.751 (22.548)	-95.09 (5.48)	-95.79 (4.68)
Peak Fz / Totalweight	-1.731 (0.514)	-1.67 (0.10)	-1.65 (0.08)
Peak Neck Flexion Moment (Nm)	9.878 (2.729)	9.30 (0.52)	8.06 (0.45)
Peak Neck Extension Moment (Nm)	-2.957 (3.148)	-1.09 (0.53)	-2.37 (0.77)

Table 11. Means (standard deviations) for dependent variables during dive to prone –  
Main Effect of NVG Condition

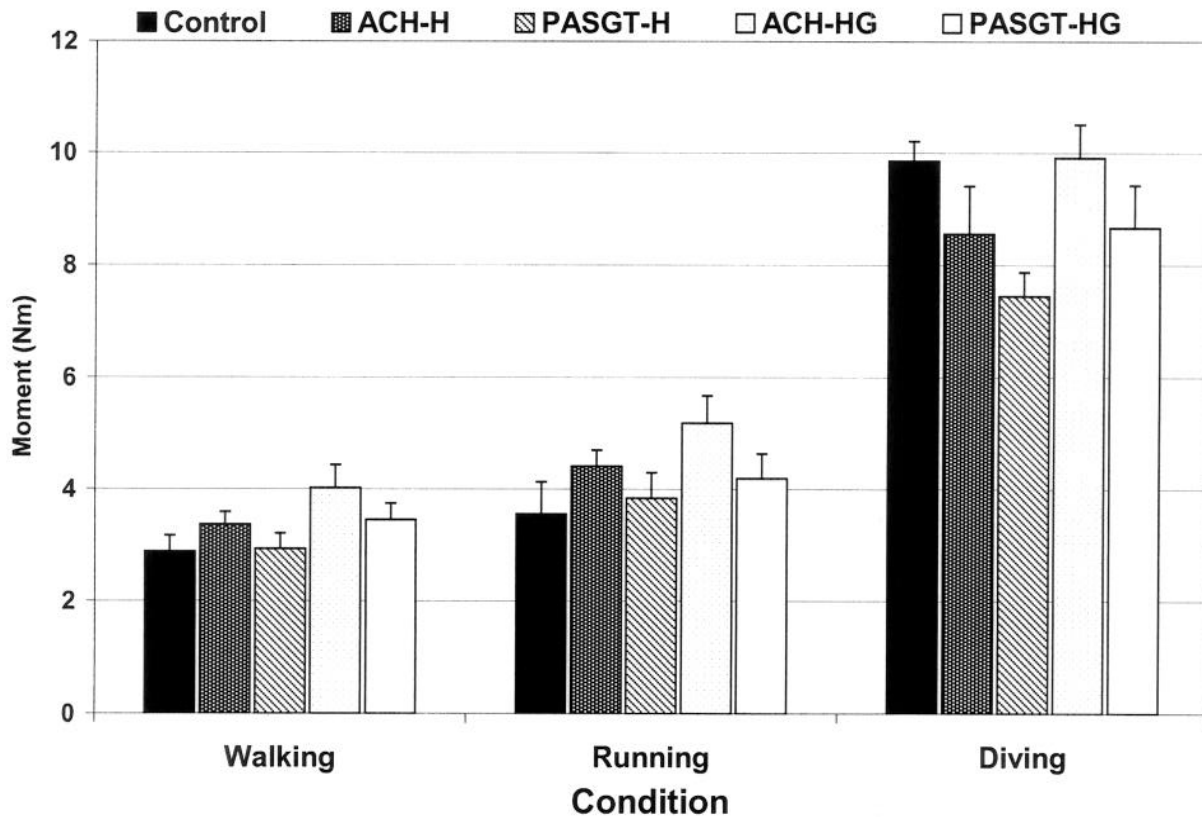
Variable	Control	Helmet	Helmet + NVG
Peak Fx Direction (N)	38.907 (24.613)	56.42 (6.22)	61.91 (6.78)
Peak Fx / Totalweight	0.843 (0.543)	1.02 (0.11)	1.03 (0.11)
Peak Fz Direction (N)	-79.751 (22.548)	-93.86 (3.82)	-97.03 (5.85)
Peak Fz / Totalweight	-1.731 (0.514)	-1.71 (0.08)	-1.61 (0.10)
Peak Neck Flexion Moment (Nm)	9.878 (2.729)	7.88 (0.43)	9.20 (0.51)
Peak Neck Extension Moment (Nm)	-2.957 (3.148)	-1.54 (0.83)	-2.14 (0.70)

Figure 14. Peak force exerted by the helmet and head on the atlas in the Z direction during each task for each helmet



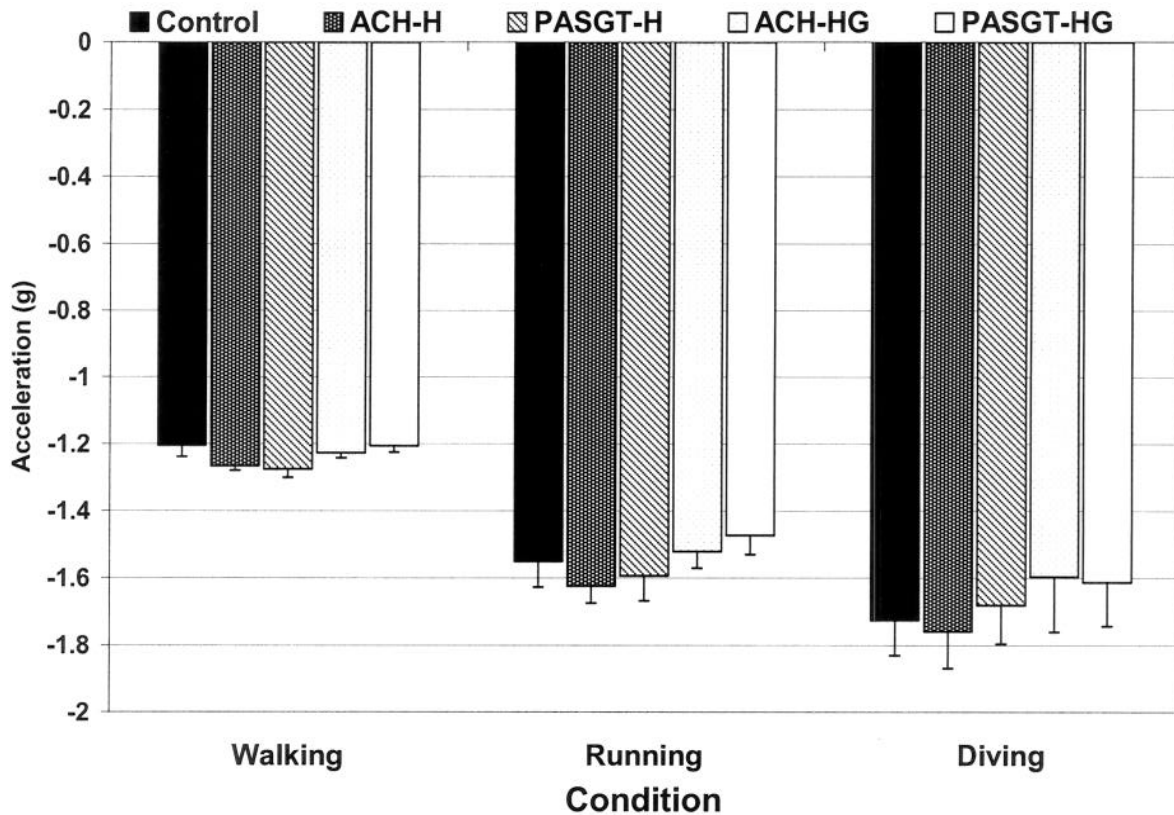
There was a significant main effect of helmet configuration (control vs. H vs. HG) on the peak neck flexion moment exerted by the helmet and head around the atlas joint during walking and diving to prone. Post-hoc analysis performed on the walking data revealed peak neck flexion moment was less in the control and H conditions than in the HG condition (Figure 15), these differences were statistically significant. For dive to prone, post-hoc analysis revealed peak neck flexion moment was less in the H condition than in the control condition, and less in the H condition than in the HG condition.

Figure 15. Peak flexion moment around the atlas attributable to the weight and acceleration of the head and helmet during each task for each helmet



Focused analysis revealed a significant main effect of NVG condition (H vs. HG) on Peak Fz during walking, on Peak Fz/totalweight (Figure 16) during walking and running, and on peak flexion and peak extension moments during dive to prone. Peak Fz during walking was greater in the HG condition than in the H condition. Conversely, Peak Fz/Totalweight while walking or running was greater in the H condition than in the HG condition. These differences were statistically significant. During dive to prone, peak flexion and extension moment were greater in the HG condition than in the H condition. Dive to Prone was the only task that was tested in which an extension moment (negative neck moment) was observed. That is, at no time during walking or running did the head/helmet exert an extension moment about the atlas. No statistically significant interactions were observed.

Figure 16. Peak Fz/Totalweight during each task for each helmet



Focused analysis also revealed a significant main effect of helmet design (ACH vs. PASGT) on peak Fz during walking and on peak flexion moment during running and diving to prone. Walking with ACH resulted in lower peak Fz than walking with PASGT. Running or diving to prone with ACH resulted in greater peak neck flexion moments than running or diving to prone with PASGT.

Regression analysis with peak compressive force exerted on the atlas as the dependent variable and weight moment as the independent variable were not statistically significant (Table 6). The relationship between peak flexion moment of the head and weight moment were statistically significant during walking and diving to prone, but not during running. However, during walking (Figure 17) and diving to prone (Figure 18) the relationship was not very strong ( $r^2 < 0.14$  and  $r^2 < 0.24$ ).

Table 6. Results from regression analysis predicting forces and moments on atlas from weight moment

Variable	R <sup>2</sup>	P-value
Walking Compressive Force on Atlas vs Weight Moment	0.0801	0.1106
Running Compressive Force on Atlas vs. Weight Moment	0.0184	0.4912
Dive to Prone Compress Force on Atlas vs Weight Moment	0.0158	0.5321
Walking Neck Flexion Moment about Atlas vs Weight Moment	0.1325	0.0373
Running Neck Flexion Moment about Atlas vs. Weight Moment	0.1051	0.0924
Dive to Prone Neck Flexion Moment about Atlas vs Weight Moment	0.2339	0.0106

Figure 17. Moment about the atlas during walking vs. weight moment

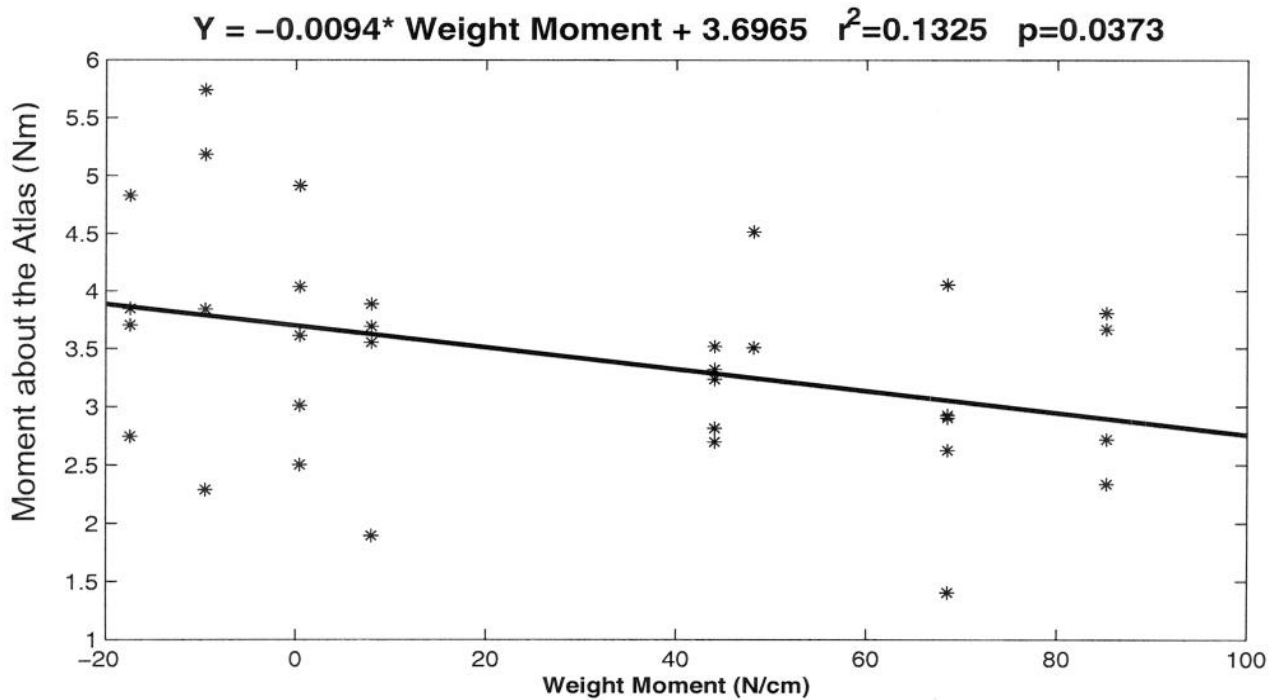
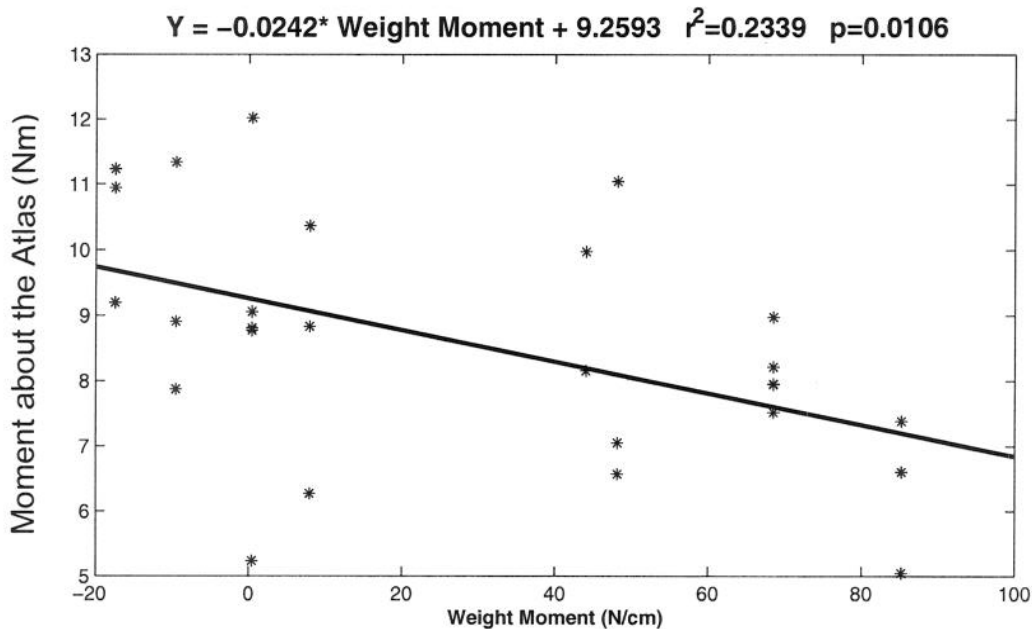


Figure 18. Moment about the atlas during dive to prone vs. weight moment



## DISCUSSION

Our first hypothesis was that wearing a ballistic helmet while walking, running and diving to prone would result in greater forces and moments on and around the atlas than wearing a lightweight control helmet. In terms of peak forces, this hypothesis was supported for all three tasks, during which neck forces were about 20-40% higher under the helmet condition than under the control condition. In terms of moments, this hypothesis was not supported for any of the three activities. In fact, diving to prone with a ballistic helmet produced lower neck flexion moments than diving to prone with the control helmet, likely due to differences in diving technique between the ballistic and control helmets.

Our second hypothesis was that adding NVG to a ballistic helmet would increase the forces and moments exerted on and around the atlas. In terms of forces, this hypothesis was supported for walking but not for running or diving to prone. In terms of moments, this hypothesis was supported for dive to prone but not for walking or running. During the dive to prone, peak neck flexion and peak neck extension moments were greater in the HG condition than in the H condition.

Our third hypothesis was that the PASGT, being of greater mass than the ACH, would produce greater forces and moments than the ACH on and around the atlas. In terms of forces, this hypothesis was supported for walking. However, no statistically significant differences in forces between the ACH and PASGT for running or diving to prone were found. In terms of moments, contrary to our hypothesis, running or diving to prone with the ACH resulted in significantly greater neck flexion moments than running

or diving to prone with the PASGT, and a similar difference during walking approached statistical significance.

Interestingly, statistically significant differences in Peak Fz/Totalweight, a dimensionless value representing acceleration in terms of gravity, were observed between the H and HG conditions during both walking and running. Peak Fz/Totalweight was less in the HG condition than in the H condition. A possible explanation for this is an adaptation in the way the soldier walks or runs when wearing a ballistic helmet with NVG that minimize head acceleration. Similar to the notion that there are adaptations in gait associated with carrying a heavy load which minimize the forces and moments associated with load carriage (LaFiandra et. al., 2002), the lower than expected peak Fz/Totalweight observed in the HG condition may reflect a mechanism that minimizes vertical force exerted by the head and helmet on the atlas.

Another interesting result was that running or diving to prone with ACH resulted in greater neck flexion moments than running or diving to prone with PASGT. Because PASGT has more mass and the COM is slightly more anterior than ACH, differences in mass properties do not explain these differences. Differences in helmet suspension type may provide an alternate explanation. One design difference between ACH and PASGT is that PASGT has a strap suspension system while the ACH has comfort pads, which are designed to increase the stability of the helmet on the head. It was observed that during running, PASGT moved 2.34° relative to the head, while ACH only moved 1.83°. Because of missing face marker data during dive to prone, helmet motion could not be calculated for that task. It may be the case that the less secure attachment of the PASGT allows the PASGT to slide slightly more relative to the head, allowing the deceleration of the helmet to occur over a greater period, resulting in a lower peak deceleration. The lower peak deceleration of the helmet thereby results in less force being applied by the helmet to the head, in turn leading to lower neck flexion moments. Additionally, it may be the case that the head actually experienced greater deceleration than the helmet (which was where deceleration was measured). The difference in deceleration between the head and helmet can not be determined from the current data collection setup.

One limitation of the current study was the assumption that the head and helmet were a single rigid body. As mentioned previously, there was some small relative movement between the head and helmet during walking and running. Thus, accelerations of the head may have been greater or smaller than those determined from the accelerometers and reflective markers on the helmet, and actual torques about the neck may have been higher or lower than those calculated. For the dive to prone, we could not determine the relative movement between the helmet and head because the face markers were not visible to the camera throughout the movement. Thus, the calculated moments for the dive to prone may have deviated even further from actual values than during walking and running. Future work in the area should be done to develop methods to analyze forces and moments on the neck exerted by headgear that avoid the assumption that the helmet and head move in conjunction. Doing so will allow for an analysis of mechanical energy transfers between the helmet and head and the effect of helmet motion on head stability during walking and running.

Previous research (Butler 1992) suggested the optimal weight moment in terms of adding mass to the helmet while minimizing the forces and moments in the neck and posterior neck muscle activity is approximately  $83 (\pm 23)$  N·cm for pilots. In actuality, the optimal weight moment would likely be near zero if it represented the head with no helmet or helmet mounted displays. Adding ballistic protection to the head and increasing the functionality of the helmet by adding night vision goggles provides advantages to the Soldier. However the advantage brought by these devices results an increase in helmet mass and non-optimal forces and moments on the neck. Thus the benefits of ballistic protection and the added functionality of head mounted displays must be balanced against the increase in injury potential and the performance decrement resulting from increased forces and moments on the neck.

It is important to note that the highest neck compression acceleration in any of these activities, 1.8 g, is less than 10% of the 20 g reported by the U.S. Air Force as the lower limit for acute neck injury (Cogswell 2006). However, that is not to say that these smaller accelerations might not cause overuse injuries. As in sports like running, in which the individual impact of each foot strike is not likely to cause traumatic injury, repetitive lower accelerations may well lead to overuse injury over time. Long term aviator exposure to 1-2 g accelerations while wearing a helmet during normal flight conditions has resulted in complaints of sore or stiff neck (McEntire 2006).

## **CONCLUSIONS**

The results from this study provide helmet developers with information relevant to the design of helmets that might reduce forces and moments on the head and neck and may concomitantly reduce neck fatigue and injury. The study also provides the basis for future work of gathering information needed to set operational guidelines for the use of head supported masses by combat foot soldiers. Future research should expand on the scope of this study to include EMG and explore relationships between neck flexion moments and anterior neck muscle activity in the combat foot soldier.

## **RECOMMENDATIONS**

1. Conduct follow-on studies to investigate the effects of helmet mass properties on neck muscle electrical activity and neck muscle fatigue. Determine if a relationship exists between helmet mass properties, neck muscle fatigue and performance of dismounted soldier tasks. Determine if physical training of the neck muscles can counteract potential fatigue effects and reduce acceleration of the head relative to the body during body jarring or acceleration.
2. A well balanced (anterior/posterior) helmet may reduce the weight moment of the helmet. However, balancing the helmet may result in an increase in moment of inertia. Research should be conducted to investigate the effects of helmet moment of inertia on neck force and moments.
3. Develop research methodology that is not based on the assumption that the helmet and head are rigidly linked.
4. Expand the variety of militarily relevant tasks analyzed.
5. Minimize helmet mass as much as possible while maintaining necessary protective properties.

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